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<b>(57) Abstract</b> <p>A catheter for emitting radiation is disclosed, comprising a catheter shaft (104), and an x-ray unit (102) attached to the distal end of the catheter shaft. The x-ray unit comprises an anode (112), and a cathode (110) coupled to an insulator (108) to define a vacuum chamber (106). The cathode is preferably a field emission cathode of graphite or graphite coated with titanium carbide, for example. The anode is preferably tungsten, and the insulator is preferably pyrolytic boron nitride. The x-ray unit is preferably coupled to a voltage source through a coaxial cable. The anode is preferably a heavy metal such as tungsten. The cathode may also be a ferroelectric material. The x-ray unit can have a diameter less than about 4mm, and a length less than about 15 mm. Methods of use of the catheter are also disclosed. The catheter of the present invention can be used to irradiate the site of an angioplasty procedure to prevent restenosis. It can also be used to treat other conditions in any vessel, lumen or cavity of the body.</p>			

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### X-RAY CATHETER

This application claims the benefit of U.S.  
5 Provisional Application Nos. 60/006,708 filed November 14,  
1995, and 60/002,722 filed August 24, 1995.

### FIELD OF THE INVENTION

The present invention relates generally to catheters  
and, more particularly, to catheters for irradiating  
10 vessels, lumens or cavities of a body, such as  
cardiovascular tissue to reduce the incidence of restenosis,  
and to treat other conditions.

### BACKGROUND OF THE INVENTION

Restenosis of an artery or vein after percutaneous  
15 transluminal coronary angioplasty (PTCA) or percutaneous  
transluminal angioplasty (PTA) occurs in about one-third of  
the procedures, requiring the procedure to be repeated.  
Various types of drugs or other agents are being  
investigated for use in preventing restenosis. Heparin, an  
20 anticoagulant and inhibitor of arterial smooth muscle  
proliferation, is one such drug. Dexamethasone may also  
prevent smooth muscle proliferation. Integralin, which  
prevents aggregation of platelets, may also be useful.  
Other anticoagulants and antiproliferative agents are being  
25 investigated for efficacy, as well. Such drugs can be  
delivered before or after the angioplasty procedure. The  
delivery of lytic agents such as urokinase, streptokinase  
and recombinant tissue type plasminogen activator (rtPA) to  
dissolve thrombi in arteries and veins is also being  
30 investigated.

Because of blood flow through the artery, drugs  
delivered to the site of an angioplasty procedure, for  
example, can be rapidly dissipated and removed from the site  
before they can be sufficiently absorbed to be effective.  
35 Catheters have therefore been developed to directly drive  
the drug into the desired site through a balloon or to  
maintain the delivered drug agent proximate the desired site  
by isolating the region with occlusion balloons. See, for

1 example, U.S. Patent Nos. 5,087,244, 4,824,436, and  
4,636,195, to Wolinsky.

5 The use of sufficient pressures to drive the drug  
into the tissue or plaque, however, may damage the arterial  
wall. Passive delivery into a region isolated by occlusions  
balloons, on the other hand, is slow and may not enable  
sufficient absorption of the medication. Passive delivery  
can be particularly inappropriate for drug delivery in an  
artery because blood flow can only be occluded in an artery  
10 for a limited period of time.

Stents have also been used after angioplasty to  
prevent an opened blood vessel from closing. The use of  
stents, however, has only shown a small decrease in the  
incidence of restenosis. Stents are also difficult to  
15 properly position and are expensive.

The use of radiation has also been investigated to  
reduce restenosis after PTCA or PTA. One technique is  
Photodynamic Therapy (PDT), wherein photosensitive drugs  
delivered to the angioplasty site are activated by  
20 irradiation with ultraviolet (UV) or visible light.

Another approach was to expose vascular tissue to UV  
light within a wavelength band of DNA absorption (240-280  
nm) by a laser to disable or destroy the DNA of the tissue.  
This would impair or destroy the ability of the vascular  
25 tissue to proliferate. This approach had only limited  
success, however, because UV light does not penetrate  
vascular tissue sufficiently to prevent proliferation or  
migration of smooth muscle tissue.

Beta-irradiation of the vessel after angioplasty  
30 with radioactive guide wires or implanted stents is another  
technique. U.S. Patent No. 5,199,939 to Dake et al., for  
example, discloses a catheter with radioactive pellets at  
its distal end to irradiate the site of an angioplasty  
procedure to prevent restenosis. The need for a radioactive  
35 source in the catheter lab, however, requires protection  
against radioactive hazards to personnel and costly  
compliance with regulations. It is also difficult to  
control the depth of penetration of the radiation by this  
method.

1 U.S. Patent No. 4,143,275 to Mallozzi et al.,  
discloses an x-ray device for delivering radiation to remote  
locations of the human body such as the interior of the  
heart. The x-ray radiation is generated by irradiating a  
5 target material, such as iron, calcium, chromium, nickel,  
aluminum, lead, tungsten or gold, by a laser to vaporize the  
metal. X-ray radiation is emitted from the ionized vapor  
plasma. The target is located outside the body and the x-  
rays are directed to a desired location within the body  
10 through a hollow guide. The patent discusses use of such a  
device to produce radiographs, to irradiate tumors or to  
alter tissue. It is believed, however, that x-ray radiation  
generated by this method would have photon energy of about  
1-2 KeV at best, which is too low to penetrate biological  
15 tissue deeper than about 20-30 microns. In addition, the  
patent does not disclose how to produce a guide which is  
both flexible enough to be advanced through the  
cardiovascular system and able to transmit adequate x-ray  
radiation to an intended site without excessive losses.

20 U.S. Patent No. 5,153,900 to Nomikos, et al.,  
discloses a miniaturized low power x-ray source for  
interstitial insertion for the treatment of tumors. The  
device comprises a housing with an elongated cylindrical,  
rigid probe. An anode and cathode are located in the  
25 housing and a target is located at the distal end of the  
probe. The cathode and target must lie along the same axis.  
Electrons emitted by the cathode, which can be a thermionic  
emitter or a photocathode, impinge on the target, causing  
the emission of x-ray radiation. A rigid probe is  
30 unsuitable for use in the cardiovascular system.

U.S. Patent No. 5,428,658 to Oettinger, et al., a  
continuation of the patent to Nomikos, discussed above,  
discloses a flexible probe comprising a flexible optical  
fiber within a metallic tube. The optical fiber has a  
35 photoemissive coating at its terminal end. A target is  
located distal to the terminal end of the optical fiber,  
within an evacuated shell. The flexible probe is said to  
enable threading down a pathway, such as the trachea, or  
around structures, such as nerves or blood vessels. Such a

1 device is not sufficiently flexible for advancement through  
the cardiovascular system, nor is it believed that such a  
device can be made small enough to access the site of a PTCA  
procedure.

5 U.S. Patent No. Re 34,421 to Parker, et al.  
discloses an x-ray microtube comprising a glass tube having  
a diameter less than one inch, for insertion into the body  
for treating a tumor. While asserting that the diameter can  
be as small as 1/8 inch, Parker does not address any of the  
10 problems associated with such a small device, such as  
electrical flashover. It is questionable whether such a  
device could be made small enough to access the site of a  
PCT procedure, and still function. Glass also has too high  
a coefficient of absorption of x-ray radiation to enable  
15 delivery of sufficient x-ray radiation to prevent restenosis  
in a reasonable period of time. Parker also does not  
disclose any way to advance its x-ray source through the  
cardiovascular system, or any other channel of the body.

#### SUMMARY OF THE INVENTION

20 In accordance with a preferred embodiment of the  
present invention, an x-ray catheter is disclosed which is  
small and flexible enough to access an intended site within  
a vascular system of the body, such as the coronary arteries  
of the cardiovascular system. The x-ray catheter can  
25 operate at the high voltages required for generating x-ray  
radiation of an effective spectrum for preventing restenosis  
and treating other conditions. It also has walls highly  
transmissive to x-ray radiation so that an effective dosage  
can be delivered in a short period of time.

30 In accordance with the present invention, a catheter  
for emitting x-ray radiation is disclosed comprising a  
flexible catheter shaft having a distal end and an x-ray  
unit coupled to the distal end. The x-ray unit comprises an  
anode, a cathode and an insulator, wherein the anode and  
35 cathode are coupled to the insulator to define a vacuum  
chamber. The insulator is preferably pyrolytic boron  
nitride, which is highly transmissive to x-ray radiation.  
The cathode is preferably a field emission cathode of  
graphite, graphite coated with titanium carbide, or other

1 carbides. The cathode can also comprise silicon and the x-ray unit can include a grid. The cathode can be a  
ferroelectric material, as well. The anode is preferably tungsten. The catheter shaft is preferably a coaxial cable.  
5 A guide wire may be provided extending through the catheter shaft, partially through the catheter shaft or partially through the x-ray unit, in a rapid exchange configuration. The catheter further preferably comprises a means for centering the x-ray unit within a lumen.

10 In accordance with another embodiment of the invention, an x-ray catheter is disclosed comprising a flexible catheter shaft for being advanced through lumens of a vascular system.

Another embodiment of the present invention  
15 comprises an x-ray generating unit having a diameter less than about 4 mm.

Yet another embodiment of the present invention comprises a catheter shaft, an x-ray generating unit and means for centering the x-ray generating unit within the  
20 lumen.

A method is also disclosed in accordance with the present invention for preventing restenosis of a lumen or treating other conditions, comprising advancing an x-ray catheter through a lumen to a first location adjacent an  
25 intended site of the lumen, wherein the x-ray catheter comprises a flexible catheter shaft with a distal end and an x-ray generating unit coupled to the distal end. The x-ray generating unit comprises an anode, a cathode and an insulator, wherein the anode and cathode are coupled to the  
30 insulator to define a vacuum chamber. The method further comprises causing the emission of an effective dose of x-ray radiation and removing the catheter. The catheter can be inserted after conducting an angioplasty procedure. The catheter can be advanced over a guide wire and through a  
35 guide catheter, or through an exchange tube.

#### DESCRIPTION OF THE FIGURES

Fig. 1A is a cross-sectional view of an x-ray catheter in accordance with a first embodiment of the present invention;

1           Fig. 1B is a cross-sectional view of a preferred catheter shaft for use in the present invention;

          Fig. 2A is a graph of an exemplary voltage applied between the anode and grid electrode versus time;

5           Fig. 2B is a graph of an exemplary voltage applied between the grid electrode and rear electrode of the cathode versus time;

          Fig. 2C is a graph of the current flow from the cathode to the anode versus time, for the voltages of Figs. 2A and 2B;

10           Fig. 2D is a graph of the power of the emitted x-ray radiation for the voltages of Figs. 2A and 2B;

          Fig. 3A is an alternative cathode in accordance with a second embodiment of the invention;

15           Fig. 3B is an enlarged cross-section of one needle of Fig. 3A;

          Fig. 4 is a graph of photon energy versus the Linear Attenuation Coefficient,  $\mu$ ;

20           Fig. 5 is a cross-sectional view of the distal portion of a third embodiment of the present invention;

          Fig. 6 is a cross-sectional view of mandrel for use in chemical vapor deposition of the insulator of the embodiment of Fig. 5;

25           Fig. 7 is a cross-sectional view of the distal portion of a fourth embodiment of the present invention;

          Fig. 8 is a cross-sectional view of the distal portion of a fifth embodiment of the present invention;

30           Figs. 9-11 are side views of the distal portions of the catheter of the present invention, including several centering devices for centering the x-ray unit within a lumen;

          Fig. 14 is a cross-sectional view of a distal portion of a catheter in accordance with the present invention, in a rapid exchange configuration wherein the guide wire passes through the distal tip of the x-ray unit; and

35           Fig. 15 is a partial cross-sectional view of another catheter in accordance with the present invention in a rapid exchange configuration wherein the guide wire enters and



1 exits the catheter shaft proximal to the x-ray unit.

#### DESCRIPTION OF THE INVENTION

Fig. 1A is a cross-sectional view of an x-ray catheter 10 in accordance with a first embodiment of the present invention. The x-ray catheter 10 comprises a flexible catheter shaft 12 adapted for insertion into blood vessels or other body vessels. The shaft 12 can be polyethylene, polyurethane, polyether block amide, nylon 12, polyamide, polyamide copolymer, polypropylene, polyester copolymer, polyvinyl difluoride or silicon rubber, for example.

A miniature x-ray unit 14 is secured at the distal end of the catheter shaft 12 by an adhesive, for example. The x-ray unit 14 comprises a vacuum chamber 16, a cathode 18, which emits electrons, and an anode 20, which receives the emitted electrons. The anode 20 abruptly decelerates the impinging electrons, causing the emission of x-ray radiation by the Bremsstrahlung effect, as is known in the art. About 0.1-0.2% of the kinetic energy of the impinging electrons is emitted in the x-ray range of about 0.5-5 Angstroms in the preferred embodiments of the present invention.

In this embodiment, the anode 20 preferably has the shape of an inverted cone. The walls of the anode 20 preferably have an angle of about 16° with respect to the surface of the cathode 18. The anode 20 is preferably a heavy metal, such as gold or tungsten, for example.

The cathode 18 comprises a base 19 which in this embodiment is preferably a ferroelectric material, as discussed below. The base 19 can also be doped or undoped silicon, or other such materials, which is also discussed below.

A grid electrode 24 is coupled to the surface of the base 19 facing the anode 20. A rear electrode 27 is coupled to the rear of the base 19. Wires 26, 28 and 30 extend from the rear electrode 27, anode 20 and the grid 24, respectively, through the catheter shaft 12, to a high voltage generator 32. The generator 32 preferably operates

1 in the 0-30 kilovolt (Kv) range. The wires 26, 28 and 30  
can be soldered in place. Separate lumens 34, 36, 38 can be  
provided through the catheter shaft 12 for each wire or a  
single lumen can be provided for a coaxial cable comprising  
5 the three wires. A coaxial cable can form the catheter  
shaft as well, as in the embodiments of Figs. 5 and 7.

The vacuum chamber 16 preferably comprises a wall 22  
of beryllium, beryllium oxide, aluminum, aluminum oxide,  
pyrolytic boron nitride, graphite or other such metal or  
10 ceramic materials, which is transparent to x-rays. If a  
metal, such as beryllium or aluminum is used as the wall 22  
of the vacuum chamber 16, an insulative layer (not shown)  
would be provided to electrically insulate the anode 20 and  
cathode 18, as is known in the art. Aluminum oxide,  
15 pyrolytic boron nitride and other ceramics are insulators.  
A transparent biocompatible coating 25 of a polymeric  
material such as polyethylene, polyurethane or Teflon (R),  
for example, is also provided over the wall 22. A vacuum  
tie off (not shown) depends from the vacuum chamber 16,  
20 which is sealed after the desired vacuum within the chamber  
is achieved. A soft, resilient material 48 may be provided  
at the distal tip of the x-ray unit 14, as is known in the  
art. The material can be ultra low density polyethylene or  
nylon, for example.

25 A lumen 40 extending longitudinally through the  
catheter shaft 12 can also be provided to accommodate a  
guide wire 42. A port 44 can be provided through the shaft  
12 for the guide wire 42 to exit the shaft 12. A tube 48  
can be attached by adhesive or thermal bonding to the shaft  
30 12 at the port 44 to provide a guide for the guide wire 42  
around the x-ray unit 14. The tube 48 may be adhered to  
the wall of the x-ray unit 14, as well. The tube 48 may  
extend through the soft material 46 at the distal tip of the  
x-ray unit 14.

35 The lumens in Fig. 1 are shown in the same plane for  
illustrative purposes. If multiple lumens are provided,  
they would preferably be arranged symmetrically within the  
catheter, as shown in Fig. 1B.

1 In this embodiment, the base 19 of the cathode 18 is  
preferably a ferroelectric material, as described in Riege,  
H., "Electron emission from ferroelectrics - a review,"  
Nuclear Instruments and Methods in Physics Research A340  
5 (1994), pp. 80-89; Gundel, H., et al., "Fast Polarization  
Changes in Ferroelectrics and Their Application," Nuclear  
Instruments and Methods in Physics Research A280 (1989), pp.  
1-6; Gundel, H., et al., "Time-dependent electron emission  
from ferroelectrics by external pulsed electric fields," J.  
10 Appl. Phys. 69(2) 15 January 1991, pp. 975-982; and Asano,  
Jun-ichi, et al., "Field-Excited Electron Emission from  
Ferroelectric Ceramic in Vacuum," Jpn. J. Appl. Phys. Vol.  
31 (1992), pp. 3098-3101, Part 1, No. 9B, which are all  
incorporated by reference herein. As described in those  
15 articles, ferroelectric materials, such as lead-zirconium-  
titanate (PZT) and lead-lanthanum-zirconium-titanate (PLZT)  
and triglycinesulfate (TGS), for example, emit electrons  
from their surfaces when the spontaneous ferroelectric  
polarization of these materials is rapidly reversed. High  
20 voltage, submicrosecond pulses can cause such reversals, as  
can mechanical pressure pulses, thermal heating or laser  
illumination. The use of a laser to cause polarization  
reversal is discussed in Geissler, K., et al., "Intense  
laser-induced self-emission of electrons from  
25 ferroelectrics," Physics Letters A 176 (1993), pp. 387-392,  
North Holland, which is also incorporated by reference  
herein. Ferroelectric cathodes do not require as high  
vacuum as other types of cathodes. A vacuum of about  $10^{-3}$ -  
 $10^{-4}$  Torr is sufficient. Ferroelectric cathodes are also  
30 simple to manufacture and are reliable.

Preferably, the polarization switching is caused by  
applying an electrical pulse across the ferroelectric  
material. Preferably, voltage pulses are applied between the  
rear electrode 27 and the grid electrode 24. Positive or  
35 negative pulses, or a combination of positive and negative  
pulses, can be used, depending on the configuration and  
original orientation of the polarization of the  
ferroelectric material. The reversal of ferroelectric  
polarization can be achieved by applying a voltage pulse of

1 between about 1-3 Kv to the ferroelectric cathode 18 via the  
rear electrode 27 and the grid electrode 24. The pulses are  
preferably applied for 5-100 nanoseconds. The polarization  
of the ferroelectric material 19 can be switched at a rate  
5 of between about 1 kHz-5 MHz. Electrical current densities  
as high as 100 Amps per square centimeter can be generated.  
With a polarization switching rate of about 100 kHz, for  
example, and a diameter of ferroelectric material 19 of  
about 1 mm, an average anode current of about 10  
10 milliamperes can be generated.

Preferably, a constant voltage or voltage pulses are  
applied between the anode and the cathode, as well, to  
control the energy of the emitted x-ray radiation, and hence  
the depth of penetration of the radiation into tissue. A  
15 voltage of about 10-30 Kv is preferred in coronary  
applications, as discussed further, below.

In this embodiment, the grid electrode 24 is  
preferably silver, aluminum or gold. About one-half of its  
area is transparent or open to electrons. The grid 24 can  
20 be deposited on a layer of ferroelectric material, such as  
PZT, PLZT or TGS, as is known in the art. The dimensions of  
the cathode 18 depend on the application. For use in  
coronary arteries, for example, the ferroelectric material  
19 can have a diameter of about 1-2 mm. For use in larger  
25 blood vessels, such as the femoral artery, the diameter of  
the ferroelectric material 19 could be up to 3 mm. The  
thickness of the ferroelectric material 19 can be between  
about 50-1,000 microns. About 200-500 microns is preferred.  
The grid 24 is preferably about 0.5-10 microns thick, with  
30 about the same diameter as the ferroelectric material 19.  
The electrode 27 is about 1 micron thick. The distance  
between the anode 20 and cathode can be about 0.2-5 mm.

Experimental data suggests that restenosis after  
PTCA can be limited by irradiation by about 2000 centigrays  
35 (cGy). (See, for example, Tim A. Fischel et al., "Low-Dose,  
beta-particle emission from "stent" wire results in  
complete, localized inhibition of smooth muscle cell  
proliferation," Circulation, Vol. 90, No. 6, December 1994,  
and Wiedermann, Joseph G., et al., "Intracoronary

1 Irradiation Markedly Reduces Neointimal Proliferation After  
Balloon Angioplasts in Swine: Persistent Benefit at 6-Month  
Follow-Up," JACC Vol. 25, No. 6, May 1995, 1451-6, which are  
incorporated by reference, herein).

5 It is believed that the x-ray unit in accordance  
with this and the other embodiments of the present invention  
disclosed herein can emit over 2000 centigrays of x-ray  
radiation in about one minute, to a cylindrical region of a  
lumen with a length of about 5 mm. Treatment of a typical  
10 lesion in a coronary artery, which can be 1-2 centimeters  
long, can require repositioning of x-ray unit several times  
to irradiate the entire lesion. A lesion 1-2 centimeters  
long can therefore be irradiated in about 2-5 minutes. The  
x-ray catheter of the present invention can deliver  
15 sufficient x-ray radiation to a lesion in a short period of  
time which minimizes the inconvenience and discomfort of the  
patient and cost of the procedure.

In operation, the high voltage generator 32  
preferably applies voltage pulses between the anode 20 and  
20 grid 24, and between the rear electrode 27 and grid 24. In  
Fig. 2A, exemplary voltage pulses applied between the anode  
20 and grid 24,  $V_{AG}$ , are plotted versus time. The voltage  
pulses in this example are about 10-12 Kv. The voltage  
pulses between the anode 20 and grid 24 can be applied for  
25 about 0.1-1.0 microseconds, every 10 microseconds. Fig. 2B  
plots exemplary voltage pulses  $V_{GR}$ , applied between the grid  
electrode 24 and the rear electrode 27 versus time. The  
voltage difference here is about 2.0 Kv. Fig. 2B also shows  
a negative pulse 49 which is preferably applied to restore  
30 the negative charge on the surface of the ferroelectric  
material 19 adjacent the grid 24. Fig. 2C illustrates  
qualitatively the current  $I_A$  flowing from the ferroelectric  
material 19 to the anode 20 for the voltage pulses shown in  
Figs. 2A and 2B. The length of each current pulse generated  
35 for the range of voltage pulses of 0.1-1 microsecond, is  
about 10-100 nanoseconds. The current pulses cause the  
emission of pulses of x-ray radiation with peak power in  
this example of up to about 30 watts, as shown in Fig. 2D.

1           In a second embodiment of the invention, shown in  
Fig. 3A, the cathode 18 may also be a field emission cathode  
50 comprising multiple needles 52 and optionally a grid  
electrode 54. Fig. 3B is an enlarged cross-sectional view  
5 of a single needle 52, of Fig. 3A. The base 55 and needles  
52 can be doped or undoped silicon. The grid 54 can be  
niobium. If a grid 54 is provided, a layer 57 of an  
insulator, such as silicon dioxide ( $\text{SiO}_2$ ), is preferably  
deposited over the base 55 of silicon. The grid 54 of  
10 niobium is deposited over the silicon dioxide layer 57. A  
rear electrode 59 is coupled to the rear of the base 55. A  
wire 58 is coupled to the rear electrode 59. A wire 56 is  
coupled to the grid 54. Returning to Fig. 3A, a vacuum tie-  
off 60 is shown, as well. The anode 20 can be the same as  
15 described above.

The radius of the tips of the needles 52 is between  
about 5-100 Angstroms. The height of the needles is about  
0.5-1.0 microns. The grid 54, which is about 0.5 microns  
thick, is preferably positioned slightly above the top of  
20 the needle 52, as shown in Fig. 3B. The openings in the  
grid 54 have a diameter of about 2 microns. The layer of  
silicon dioxide is about 1-2 microns thick. A vacuum of  
between about  $10^{-7}$ - $10^{-8}$  Torr is preferred for a field  
emitting cathode including silicon.

25           The needles 52 emit electrons when negative  
potential is applied between the rear electrode 59 and the  
grid electrode 54. A triggering voltage of about 100-500  
volts may be used, for example. The voltage can be constant  
or pulsed. If no grid electrode is provided, the high  
30 voltage can be provided directly between the anode and the  
needles 52.

The radiation emitted by the anode 18 passes through  
the vacuum chamber wall 22 and coating 25, into surrounding  
tissue. Irradiation reduces the ability of smooth muscle  
35 cell to proliferate, inhibiting restenosis, as discussed  
above. Fig. 4 is a graph of Photon Energy (kev) versus the  
Linear Attenuation Coefficient  $\mu$  ( $\text{cm}^{-1}$ ) for bone 62, muscle  
64 and lung tissue 66. (See, Anthony Brinton Wolbarst,  
Physics of Radiology, Appleton and Lange, 1993, p. 108;

1 Johns, H.E., Cunningham, JR.: The Physics of Radiology, 4th  
ed., Springfield, IL; Charles C. Thomas, 1983, Appendix A.)  
The greater the coefficient  $\mu$ , the more effectively the  
medium absorbs and scatters photons. The depth of  
5 penetration of radiation is the depth at which the intensity  
of the impinging radiation drops to  $1/e$  of its original  
value. The depth of penetration of x-ray radiation of a  
particular energy is equal to  $1/\mu$ . Generally, the  
coefficient  $\mu$  increases with increasing effective atomic  
10 number of the material. While muscle and lung tissue have  
nearly identical chemical composition, the attenuation in  
muscle tissue is about 3 times greater than the attenuation  
in lung tissue, because muscle tissue is about 3 times  
denser than lung tissue. The energy of x-ray radiation is  
15 preferably adjusted so that it penetrates slightly into the  
adventitia tissue of the blood vessel about 2 mm deep.  
Penetration into the cardiac muscle tissue beyond the  
coronary artery, for example, should be minimized. The  
energy can be adjusted by varying the voltage applied  
20 between the anode and cathode. The preferred voltage range  
of 10-30 Kv yields x-ray radiation with a peak energy of  
about 8-10 KeV, which is appropriate in coronary  
applications.

Fig. 5 is a cross-sectional view of the distal  
25 portion of an x-ray catheter 100 in accordance with a third  
embodiment of the present invention. The x-ray catheter 100  
comprises an x-ray unit 102 coupled to a high voltage  
coaxial cable 104. The x-ray unit 102 has a vacuum chamber  
106, defined by an insulator 108, a cathode 110 and an anode  
30 112. The insulator 108 comprises a base portion 114 coupled  
to a tubular, preferably cylindrical wall portion 116 with  
an open end 118. The cathode 110, which is a cold, field  
emission cathode, is coupled to the open end 108. The  
insulator 108 is preferably alumina, beryllium oxide or more  
35 preferably, pyrolytic boron nitride. The boron nitride must  
be pyrolytic, as opposed to sintered, because only the  
pyrolytic boron nitride is vacuum tight at the wall  
thicknesses required. The cathode 110 is preferably  
graphite. The anode 112 is preferably tungsten or tungsten

1 coated with a layer of platinum. A one micron layer of  
platinum is sufficient. The vacuum is preferably  $10^{-5}$  Torr  
or better.

5 The cathode 110 is preferably graphite, carbides,  
such as titanium carbide, silicone, metals, or graphite  
coated with titanium carbide. The cathode 110 preferably  
includes one or a plurality of protrusions 110a with a sharp  
tip extending towards the anode 112 along a central axis of  
10 the x-ray unit 102. The protrusion 110a locally enhances  
the electrical field and improves the emission of electrons,  
as is known in the art. The protrusion 110a can comprise  
the same material as the cathode 110, or can be another of  
the cathode materials mentioned above.

15 The anode 112, which is preferably in the shape of a  
rod, extends along the central axis of the x-ray unit 102.  
The rod 112 has a depending portion 112a received within a  
cylindrical groove 114a extending through the base portion  
114. Preferably, the base 114 has a portion 114b, which  
tapers toward the anode 112. An angle of about  $45^\circ$  can be  
20 used, for example. The anode 112 also can have a portion  
112b tapered toward the cylindrical portion 114b of the  
base. Such a configuration displaces the electrical field  
from the anode-vacuum-insulator triple junction, decreasing  
the risk of electrical flashover during operation. The  
25 anode 112 is preferably a heavy metal. Tungsten is  
preferred.

The cathode 110 and anode 114 are coupled to the  
high voltage generator 32 of Fig. 1, described above,  
through the high voltage coaxial cable 104. The coaxial  
30 cable 104 comprises a central conductor 120, which is  
coupled to a proximal end of the anode 114, and an external  
conductor 122, which is coupled to the cathode 110. A  
conductive coating 124 is provided over the external surface  
of a portion of the cathode 110 and the external surface of  
35 the insulator 108 to couple the cathode 110 to the external  
conductor 122. A silver coating with a thickness of about  
0.1-1.0 microns may be used. Gold may be used as well.  
Insulation 126, such as Teflon (R), silicone, rubber,  
fluorinated ethylene propylene (FEP) or polyethylene, for



1 example, is typically provided between the external  
conductor 122 and the central conductor 120. The x-ray unit  
102 can be attached to the coaxial cable 114 with an  
adhesive, for example.

5 The cathode's "triple junction point" (the junction  
between the cathode, the insulator and the vacuum), which in  
this embodiment is an annular region surrounding the cathode  
110 proximate the open end 118 of the insulator 108, is  
screened from the high electrical field between the anode  
10 112 and the cathode 110 by the conductive coating 124 and  
the side of the cathode 110. This decreases the incidence  
of electrical flashover, enabling the use of higher  
voltages.

The cathode 110 can be coupled to the open end 118  
15 of the insulator 108 through a metal ring 130. The metal  
ring can comprise tungsten, platinum, or graphite covered by  
platinum. Coupling of the cathode 110 to the metal ring and  
coupling of the anode 112 to the insulator 108 is described  
further, below.

20 A biocompatible layer 128 is provided over the  
external conductor 116, conductive layer 124, and the  
cathode 110. A thickness of less than about 0.002 inches is  
preferred. Preferably, the biocompatible coating 128 also  
acts as an insulating layer. The biocompatible coating may  
25 be silicone or FEP, for example. A lubricious layer (not  
shown) of a hyaluronic coating, for example, may be provided  
as well. The biocompatible coating may have sufficient  
lubricity without a further coating. Silicone, for example,  
is a highly lubricious biocompatible coating.

30 The coaxial cable 104 is chosen to have sufficient  
flexibility to be advanced through the cardiovascular or  
other such system, to an intended site. It has been found  
that standard high voltage coaxial cables are generally not  
flexible enough to be advanced through the cardiovascular  
35 system to the coronary arteries. It has further been found,  
however, that miniature high frequency coaxial cables are  
available with sufficiently small diameter (about 1.0-3.0 mm  
outer diameter) and sufficient flexibility to be advanced to  
the coronary arteries. Usually, such cables are used in

1 high frequency applications at voltages less than several  
kilovolts. Surprisingly, it has been found in connection  
with the present invention, that these cables can hold  
direct current voltages as high as 75-100 Kv without  
5 breakdown, and consequently can be used with the x-ray unit  
of the present invention for operational voltages of up to  
30-40 Kv. Such voltages are sufficient to generate x-ray  
radiation in appropriate energy ranges for the treatment of  
restenosis and other conditions. Suitable coaxial cables  
10 include CW2040-3050FR; CW2040-30; CW2040-3675-SR; and  
CW2040-3275SR, distributed by Cooner Wire, Inc. Chatsworth,  
CA, for example. Cooner distributes coaxial cables for New  
England Electric Wire Corporation, Lisborn, New Hampshire.

An x-ray unit 102 in accordance with this embodiment  
15 of the invention can have a length less than about 15 mm and  
a diameter less than about 4.0 mm, depending on the  
application. The distance between the cathode 108 and the  
anode 110 can be between about 2.0-0.2 mm, depending on the  
size of the x-ray unit 102. The thickness of the  
20 cylindrical insulator wall 116 can be between about 0.2-0.5  
mm. The diameter of the coaxial cable 104 can be about the  
same as the diameter of the x-ray unit 102. For use in  
preventing restenosis after dilatation of a coronary artery,  
which typically has a diameter of about 3 mm, the x-ray unit  
25 102 preferably has a length of about 7 mm and a diameter of  
about 1.5 mm. In peripheral blood vessels, which are  
larger, the x-ray unit 102 preferably has a diameter of  
about 3.5 mm and a length of between about 7-15 mm. Larger  
x-ray units with greater diameters and lengths than those  
30 discussed above could also be made and used in accordance  
with the present invention.

To operate the x-ray unit 101 to prevent restenosis  
in a vessel of the cardiovascular system, for example,  
direct current having a voltage of between about 10-30 Kv,  
35 can be applied to the central conductor 120. The external  
conductor is connected to ground. Electrons emitted from  
the cathode 110 due to a field emission effect impact the  
anode 112, causing the emission of x-ray radiation of about  
8-10 KeV, as discussed above. The radiation is primarily

1 emitted radially, to the vessel wall. About 10-30 Kv is  
preferred for use in the prevention of restenosis. Higher  
voltages will cause the emission of x-ray radiation of  
higher energy which can penetrate too deeply into the vessel  
5 wall, damaging cardiac tissue. Higher voltages may be used  
for other applications.

Voltages at the higher end of the 10-30 Kv range are  
preferred because the use of higher voltages enables the  
generation of the same amount of radiation with less current  
10 than the use of a lower voltages, and is therefore more  
efficient. Higher voltages also enable the generation of x-  
ray radiation of higher power. Higher power, however, can  
cause the generation of more heat, which can damage the  
tissue of a vessel wall. In this embodiment, most of the  
15 heat is generated at the anode 110 positioned at the center  
of the x-ray unit, as far from the vessel wall as possible.

Higher voltage also increases the risk of electrical  
flashover at the anode and cathode triple junctions. As  
discussed above, the anode 112 and cathode 110 are  
20 preferably configured to minimize the risk of flashover.

Bulk electrical breakdown is also a risk with  
increased voltages. Pyrolytic boron nitride has a high  
dielectric strength, enabling the x-ray unit of the catheter  
to tolerate the voltages used in this application without  
25 bulk electrical breakdown. The dielectric strength of  
pyrolytic boron nitride is 200-600 KV/mm.

Pyrolytic boron nitride is also particularly  
preferred as the insulator 108 because it is highly  
transparent to soft x-rays and can therefore be efficiently  
30 used as an x-ray window. The coefficient of linear  
absorption of boron nitride at about 8 Kev, the average  
energy of the emitted radiation, is  $1.0 \text{ mm}^{-1}$ . About 8-10  
KeV is the preferred energy level of x-ray radiation in the  
treatment of restenosis, as discussed above. Transmission  
35 of radiation through pyrolytic boron nitride with a  
thickness of about 0.3 mm is about 70%. This enables  
irradiation of tissue at a rate of at least about 1 gray per  
minute. Preferably, about 10-30 grays per minute of  
radiation at about 8-10 KeV are provided, enabling delivery

1 of an effective amount of radiation to prevent restenosis to  
a lesion about 5 mm long in about 1 minute. It is believed  
that x-ray radiation can be delivered at a rate of up to  
about 100 grays per minute with the x-ray unit of this  
5 embodiment. A lesion 1-2 cm long can be treated in about 2-  
5 minutes by progressively repositioning the x-ray unit to  
irradiate additional portions of the lesion.

Positive electrical pulses with a peak voltage of  
between about 15-30 Kv and 2-100 nanoseconds long can also  
10 be applied to the central conductor 120 of the coaxial cable  
104 at a rate of between about 1-50 KHz. The high voltage  
pulses cause field emission. The pulses can further cause a  
vacuum electrical breakdown, causing electrons to flow from  
the cathode 110 to the anode 112 through a plasma of  
15 vaporized cathode and anode material between the cathode 110  
and the anode 112.

The anode 114 is preferably attached to the  
insulator 108 of pyrolytic boron nitride during formation of  
the insulator 108 by chemical vapor deposition (CVD).  
20 During CVD, the deposited boron nitride chemically bonds to  
the anode material, forming a strong, vacuum tight seal.  
The seal formed by CVD has higher voltage hold-off because  
it does not have voids which can locally enhance the  
electrical field and cause electrical flashover.

25 A mandrel 250 for use in manufacturing the x-ray  
unit 102 by CVD is shown in Fig. 6. The mandrel 250 is  
preferably graphite. A cavity 252 is provided in the  
mandrel 250 for receiving the anode 114. The anode 114 is  
secured in an anode holder 254 of boron nitride, for  
30 example. The mandrel 250 includes a shoulder 254 for  
supporting the metal ring 130. The metal ring 210 is held  
in place by a cylindrical ring holder 256, also of boron  
nitride, for example, which is supported by a mandrel holder  
258 of graphite, for example.

35 The assembly of Fig. 6 is placed in a CVD reactor  
for the deposition of boron nitride by CVD, as is known in  
the art. Chemical vapor deposition of boron nitride is

1 described, for example, in Matsuda, et al., "Synthesis and  
Structure of Chemically Vapour-Deposited Boron Nitride,"  
Journal of Materials Science 21 (1986) pp. 649-658; and  
Pouch, John J., et al. "Synthesis Properties of Boron  
5 Nitride," Materials Science Forum, Volumes 54 and 55 (1990)  
pp. 141-152, for example, which are incorporated by  
reference, herein. The boron nitride is deposited on the  
hot surface of the assembly, crystallizing into a hexagonal  
structure. CVD of pyrolytic boron nitride can be performed  
10 by CVD Products Incorporated, of Hudson, New Hampshire, for  
example.

It may be advantageous to deposit and impregnate  
boron onto the surface of the graphite mandrel 250 and  
tungsten anode 114 prior to depositing the boron nitride.  
15 To increase the chemical stability of the anode 114 during  
the deposition procedure, the tungsten could be coated with  
a layer of platinum about 1 micron thick.

After completion of the CVD process, the mandrel 250  
is removed from the assembly by oxidation of the graphite,  
20 also as known in the art.

The cathode 110 is then vacuum brazed to the metal  
ring 130 with brazing materials, which are discussed below,  
sealing the chamber. Vacuum brazing is also known in the  
art and can be provided by Koral Labs., Minneapolis, St.  
25 Paul, for example. The sealed chamber is then covered with  
the conductive coating 124 by metal vapor deposition, for  
example.

Such a process can be used for mass production of  
large numbers of assemblies.

30 A fourth embodiment of an x-ray unit 300 in  
accordance with the present invention is shown in Fig. 7.  
The x-ray unit 300 comprises a vacuum chamber 302 defined by  
an insulator 304, preferably of pyrolytic boron nitride, a  
cathode 306, and an anode 308. The anode 308 is preferably  
35 tungsten.

The cathode 306 may be graphite, titanium carbide,  
graphite coated with titanium carbide or stainless steel,  
for example. Graphite coated with titanium carbide is  
preferred. A coating of several microns may be used.

1 Titanium coating can be provided by Lanxide Coated Products,  
Inc., Newark, Delaware, for example. The cathode 306  
preferably includes an annular protrusion 306c for creating  
a cavity for containing the brazing material 316. The  
5 cathode 306 may also include a protrusion 306a directed  
towards the anode 308, as in the embodiment of Fig. 5.

The insulator 304 comprises a cylindrical wall 304a  
with an inclined depending wall 310 and a cylindrical wall  
314 preferably parallel to the cylindrical wall 304a. The  
10 depending wall 310 is preferably angled towards the interior  
of the vacuum chamber 302. The cylindrical wall 314 defines  
a sleeve for receiving a depending portion 318 of the anode  
308. The anode 308 is coupled to the cylindrical wall 314  
through a brazing alloy 312. The cathode 306 is coupled to  
15 the open end 314 of the insulator 304 through a brazing  
alloy 316, as well.

The depending portion 318 of the anode 308  
preferably includes a slot 320 for receiving the central  
conductor 322 of a coaxial cable 324. The cathode 306 is  
20 coupled to the external conductor 326 of the coaxial cable  
324 through a conductive layer 325, as in the embodiment of  
Fig. 5. A biocompatible coating is also provided over the  
coaxial cable 324, conductive layer 325 and cathode 306. A  
lubricious coating (not shown) may be provided, as well.

25 Preformed pyrolytic boron nitride of the desired  
sizes and shapes is available from CVD Products,  
Incorporated, for example.

Appropriate brazing alloys for coupling pyrolytic  
boron nitride to the tungsten anode 308 include Incusil-15  
30 ABA and Incusil-ABA, for example, available from GTE  
Products Corporation, WESTGO Division, Belmont, C.A.  
("WESTGO"). Incusil-15 ABA comprises 14.5% indium, 1.25%  
titanium, 23.5% copper and 60.75% silver. Incusil-ABA  
comprises 12.5% indium, 1.25% titanium, 27.5% copper and 59%  
35 silver. The brazing temperatures for both alloys is about  
750°C. The brazing material can be in the form of a  
cylindrical ring placed within the sleeve formed by the  
cylindrical wall 314 in Fig. 7. The brazing material  
spreads into the vertical region between the anode 308 and

1 wall 314 during the brazing process. These alloys can also  
be used to braze the cathode 110 to the metal ring 130 in  
the embodiment of Fig. 5.

5 Appropriate brazing alloys for coupling a cathode  
308 of graphite or graphite coated with titanium carbide to  
pyrolytic boron nitride include Cusin-1 ABA and Cusil-ABA,  
also available from WESTGO. Cusin-1 ABA comprises 34.25%  
copper, 1.75% titanium, 1.0% tin and 63% silver. Cusil-ABA  
10 comprises 63% silver, 35.25% copper and 1.75% titanium. The  
brazing temperatures for both alloys is about 850°C. The  
brazing is also conducted in a vacuum of about  $10^{-5}$  Torr or  
better. Because it requires a higher brazing temperature,  
the graphite cathode 306 is coupled to the pyrolytic boron  
nitride prior to the tungsten anode 308. The brazing  
15 material can be in the form of a ring or it can be sputtered  
onto the end of the pyrolytic boron nitride prior to vacuum  
brazing.

Instead of a cathode of graphite, the cathode can be  
PLZT or other such ferroelectric material, as discussed  
20 above. As above, the use of ferroelectric material requires  
the use of voltage pulses. In Fig. 8, a fifth embodiment of  
the present invention is shown, comprising a ferroelectric  
cathode 130 supported by a conductive cap 132. The  
conductive cap 132 is coupled to the outer conductor 116 of  
25 the coaxial cable 114 by a conductive layer 118, as above.  
The remainder of the x-ray catheter 150 is the same as the  
embodiment of Fig. 5. Graphite is preferred as the  
conducting material because it has a low absorption  
coefficient for x-ray, enabling transmission through the  
30 distal end of the x-ray unit.

It is preferable to center the x-ray unit within the  
vessel or lumen, to provide a uniform distribution of x-ray  
radiation around the circumference of the vessel wall. Fig.  
9 is a side view of an x-ray catheter 400 in accordance with  
35 the present invention, with a centering device comprising a  
plastic sleeve 402 with a plurality of resilient polymeric  
solid arms 404 depending from it at an angle. The sleeve  
402 can be coupled to the outer, biocompatible layer of the  
coaxial cable 406 proximal to the x-ray unit 408 by adhesive

1 or thermal bonding, for example. The distal ends of the  
arms 404 can optionally extend beyond the distal end of the  
x-ray unit 408. The arms 404 bear against the vessel wall  
410, centering the x-ray unit 408 within a vessel or lumen  
5 of the body.

A sheath 412 is preferably provided over the coaxial  
cable 406 for compressing the arms 404 during advancement of  
the x-ray unit 408 to the intended site. When the x-ray  
unit 408 is properly positioned, the sheath 412 is  
10 retracted, releasing the arms 404. Radiopaque bands 414 of  
gold or tantalum, for example, are preferably provided on  
the coaxial cable 406 and the sheath 412 to assist in  
tracking of the x-ray catheter 400 on a fluoroscope during a  
procedure. The bands 414 are preferably positioned on the  
15 coaxial cable 406 and the sheath 412 such that when the  
sheath 412 has been sufficiently retracted to release the  
arms 404, the bands on the coaxial cable 406 and the sheath  
412 are essentially aligned.

Fig. 10 is a partial, cross-sectional view of the x-  
20 ray catheter 400 of Fig. 9, wherein the x-ray unit 408 is  
within the sheath 412 and the arms 404 are compressed.  
Saline or some other cooling agent can be delivered through  
the space 416 between the sheath 412 and the coaxial cable  
406, as well.

25 Alternatively, a compressible cage 418 can be  
provided over the x-ray unit 408 as a centering device, as  
shown in Fig. 11. The cage 418 comprises a plurality of  
arms 420 with a first end 420a coupled to a first sleeve  
portion 422 and a second end 420b coupled to a second sleeve  
30 portion 424. The x-ray catheter unit 408 extends into and  
lies within the region defined by the arms 418. The arms  
408 can be compressed by the sheath 412, as in Fig. 14. The  
second portion 424 can be coupled to the distal end of the  
x-ray unit 308.

35 The material of the outer layer of the coaxial cable  
406 and the material of the sheath 412 preferably comprise  
materials which slide easily with respect to each other.  
The outer layer of the coaxial cable 406 is preferably



1 coated with a lubricious material, such as silicone or a  
hyaluronic coating, as well.

Releasable arms and cages, methods of their  
manufacture and suitable materials are disclosed in U.S.S.N.  
5 08/488,216, filed on June 7, 1995 and assigned to the .pa  
assignee of the present inventor. U.S.S.N. 08/488,216 is  
incorporated by reference, herein.

Another method of centering the x-ray unit is a  
malecot device, as shown in Figs. 12-13. A sheath 450 of  
10 plastic material is attached to the distal portion 454a of  
an x-ray unit 454, which is shown in Fig. 12. The coaxial  
cable 456 attached to the proximal end of the x-ray unit, is  
also shown in phantom. A plurality of lateral slots 457 are  
provided through portions of the sheath surrounding the x-  
15 ray unit 454. Four equidistantly positioned slots 457 may  
be provided around the circumference of the sheath 450, two  
of which are shown in Fig. 12. The length of the slots 457  
depends on the diameter of the vessel at the intended site  
and the diameter of the sheath 450, and should be sufficient  
20 to enable the buckled portion of the sheath 450 to bear  
against the circumference of the vessel wall. When the x-  
ray unit 454 is adjacent the intended site, the sheath 450  
is advanced, causing a portion 458 of the sheath 450 between  
the slots 457 to buckle outward, as shown in Fig. 13. The  
25 sheath 450 is advanced a sufficient distance for the portion  
458 to buckle sufficiently to bear against the vessel wall,  
centering the x-ray unit 454. The distal tip 460 of the  
catheter may be of a soft, resilient material such as ultra  
low density polyethylene or nylon, for example, as is known  
30 in the art. Any of the embodiments of the x-ray catheter  
can be provided with a soft tip.

The x-ray unit could also be placed within an  
expandable balloon.

The x-ray catheters of the embodiments of Figs. 5, 7  
35 and 8 can be conveyed to the site of the dilatation  
procedure through an exchange tube after the dilatation  
catheter is removed. The exchange tube can be advanced to  
the intended site over the same guide wire used in the  
dilatation procedure. After the exchange tube is properly

1 positioned, the x-ray catheters of Figs. 5, 7 and 8 can be  
advanced through the exchange tube, to the intended site.

The x-ray catheter of the present invention can also  
be advanced over the same guide wire used by the dilatation  
5 catheter after the dilatation catheter is removed, through a  
guide catheter. Fig. 1 shows one such x-ray catheter 10.  
Fig. 14 is a cross-sectional view of another x-ray catheter  
500 for use with a guide wire 502 in a rapid exchange  
configuration. The guide wire 502 enters the x-ray unit 504  
10 through an opening 506 in the cylindrical wall of the unit  
404, extends through the center of the unit 504 and a  
central passage 508 in a cathode 510, exiting through an  
opening at the distal end of the unit 504.

The cathode 510 of the x-ray unit 504 may be  
15 graphite, for example. The anode can comprise a base 514 of  
tungsten, for example, with a plurality of rod-like  
protrusions 516 arranged concentrically about the base,  
within a vacuum cavity 518 defined by an insulator 520 and a  
cathode 510. The protrusions 516 extend toward the cathode  
20 510. The insulator 520 is preferably of pyrolytic boron  
nitride. A tube 522 of insulative, vacuum tight material,  
may be provided through the vacuum chamber 518, providing a  
passage for the guide wire 502.

The base 514 of the anode has a depending portion  
25 514a, preferably coupled to the central electrode 417 of a  
coaxial cable 518. A conductive layer is provided over the  
outer walls of the insulator 520, to couple the cathode 510  
to the outer electrode of the coaxial cable 518, as  
described in the embodiments, above.

30 Fig. 15 is a side view of another embodiment of a  
rapid exchange x-ray catheter 600 in accordance with the  
present invention, wherein a portion of the catheter shaft  
602 is shown in cross-section. Here, a lumen 601 is  
provided in the catheter shaft 602 with an entrance port 603  
35 and an exit port 604 proximal to the x-ray unit 605. A  
guide wire 606 enters the lumen 601 through a port 603 and  
exits through a port 604. The x-ray catheter 600 can be  
tracked along the guide wire 606 to the intended site in a  
lumen or vessel, through the lumen 601. The distance

1 between the entrance port 602 and the exit port 604 can be  
about 10-20 cm, for example. Other lumens (not shown) can  
be provided for a coaxial cable or wires to couple the x-ray  
unit 605 to the high voltage generator 32 shown in Fig. 1,  
5 for example.

Such a catheter shaft 602 can be formed in a multi-  
lumen extrusion process, as is known in the art, wherein the  
lumens extend longitudinally through the catheter shaft 602.  
The portions of the lumen distal and proximal to the  
10 intended locations of the exit port 604 and entrance port  
602 can be closed, as is known in the art. The ports 603,  
604 can then be made through the catheter shaft by a laser,  
for example.

While the above embodiments are described with  
15 respect to applying x-ray radiation to the site of an  
angioplasty procedure, the present invention can be used to  
apply radiation within the cardiovascular system for other  
purposes, or to other vessels, lumens, or cavities in the  
body, wherever the application of radiation would be useful.

20 The various embodiments set forth above are for the  
purpose of illustration. It will be appreciated by those  
skilled in the art that various changes and modifications  
may be made to these embodiments without departing from the  
spirit and scope of the invention as defined by the claims,  
25 below.

30

35

1 We claim:

2. A catheter for emitting x-ray radiation  
comprising:

5 a flexible catheter shaft having a distal end;  
an x-ray unit coupled to the distal end, wherein  
the x-ray unit comprises an anode, a cathode and an  
insulator, wherein the anode and cathode are coupled to the  
insulator to define a vacuum chamber.

10 2. The catheter of claim 1, wherein the cathode is  
a field emission cathode.

3. The catheter of claim 1, wherein the catheter  
shaft comprises a coaxial cable.

15 4. The catheter of claim 1, wherein the insulator  
is chosen from the group consisting of beryllium oxide,  
aluminum oxide, or pyrolytic boron nitride.

5. The catheter of claim 1, wherein the cathode and  
the anode are coupled to a voltage generator.

6. The catheter of claim 1, further comprising a  
guide wire lumen.

20 7. The catheter of claim 6, wherein the guide wire  
lumen extends partially through the catheter shaft.

8. The catheter of claim 6, wherein the guide wire  
lumen extends partially through the x-ray unit.

25 9. The catheter of claim 1, further comprising a  
means for centering the x-ray unit within a lumen.

10. The catheter of claim 1, wherein the cathode is  
a ferroelectric material.

11. An x-ray catheter comprising:

30 a flexible catheter shaft for being advanced  
through lumens of the vascular system, the catheter shaft  
having a distal end;

35 an x-ray unit coupled to the distal end, the x-  
ray unit comprising an anode, a cathode and an insulator,  
wherein the anode and cathode are coupled to the insulator  
to define a vacuum chamber.

12. The catheter of claim 11, wherein the insulator  
comprises pyrolytic boron nitride.

1           13. The catheter of claim 11, wherein the anode  
comprises tungsten or platinum and the cathode comprises  
graphite.

5           14. The catheter of claim 11, wherein the cathode  
is a field emission cathode.

          15. The catheter of claim 12, wherein the cathode  
and anode are coupled to a voltage generator.

10          16. The catheter of claim 15, wherein the catheter  
shaft comprises a coaxial cable coupling the anode and  
cathode to the voltage generator.

          17. The catheter of claim 16, further comprising  
means for centering the x-ray unit within a lumen.

          18. A catheter for the emission of x-ray radiation  
comprising:

15               a flexible catheter shaft having a distal end;  
                  an x-ray generating unit coupled to the distal  
end, the x-ray generating unit comprising an anode, a  
cathode and an insulator, wherein the anode and cathode are  
coupled to the insulator to define a vacuum chamber, and  
20               wherein the x-ray generating unit has a  
diameter less than about 4 mm.

          19. The catheter of claim 18, wherein the x-ray  
generating unit has a diameter of about 1 mm.

25          20. The catheter of claim 19, wherein the x-ray  
generating unit has a length of about 7 mm.

          21. The catheter of claim 18, wherein the x-ray  
generating unit has a length less than about 15 mm.

          22. The catheter of claim 18, wherein the insulator  
comprises pyrolytic boron nitride.

30          23. An x-ray catheter for use in irradiating the  
wall of a lumen comprising:

                  a flexible catheter shaft having a distal end;  
                  an x-ray generating unit; and  
                  means for centering the x-ray generating unit  
35          within the lumen.

          24. A method for preventing restenosis of a lumen  
comprising:

                  (a) advancing an x-ray catheter through a  
lumen to a first location adjacent an intended site of the

1 lumen, wherein the x-ray catheter comprises a flexible  
catheter shaft with a distal end and an x-ray generating  
unit coupled to the distal end, the x-ray generating unit  
comprising an anode, a cathode and an insulator, wherein the  
5 anode and cathode are coupled to the insulator to define a  
vacuum chamber;

(b) causing the emission of an effective dose  
of x-ray radiation to prevent restenosis; and

(c) removing the catheter.

10 25. The method of claim 24, wherein step (b)  
comprises causing the emission of radiation within a  
particular energy range to achieve a particular depth of  
penetration.

15 26. The method of claim 24, wherein the causing  
step (b) further comprises applying a predetermined voltage  
between the anode and the cathode to achieve the particular  
depth penetration.

27. The method of claim 24, further comprising  
irradiating tissue at a rate of 1-100 grays per minute.

20 28. The method of claim 27, wherein the irradiating  
step is conducted for about 1 minute.

29. The method of claim 24, wherein step (b)  
comprises causing the emission of x-rays having an energy of  
about 8-10 KeV.

25 30. The method of claim 24, further comprising  
centering the x-ray unit within the lumen prior to the step  
(b).

30 31. The method of claim 24, wherein the advancing  
step comprises advancing the x-ray catheter through a lumen  
of the vascular system through an exchange tube.

32. The method of claim 24, wherein the advancing  
step comprises advancing the x-ray catheter through a lumen  
of the vascular system over a guide wire and through a guide  
catheter.

35 33. The method of claim 32, wherein a portion of  
the x-ray catheter is advanced over the guide wire.

34. The method of claim 24, further comprising  
positioning the x-ray unit at a second location and causing  
the emission of x-ray radiation at the second location.

1           35. The method of claim 24, further comprising  
positioning the x-ray unit at a plurality of locations and  
causing the emission of x-ray radiation at each of the  
plurality of locations.

5           36. The method of claim 24, further comprising  
conducting an angioplasty procedure prior to step (a),  
wherein the intended site of step (a) is the site of the  
angioplasty procedure.

10          37. A method for providing x-ray radiation  
treatment comprising:

            advancing an x-ray catheter through a lumen to  
an intended site, wherein the x-ray unit comprises a  
flexible catheter shaft with a distal end and an x-ray  
generating unit coupled to the distal end, the x-ray  
15      generating unit comprising an anode, a cathode and an  
insulator, wherein the anode and cathode are coupled to the  
insulator to define a vacuum chamber;

            causing the emission of an effective dose of x-  
ray radiation; and  
20           removing the catheter.

            38. The catheter of claim 2, wherein the cathode is  
chosen from the group consisting of graphite, titanium  
carbide, carbides, metals, and graphite coated with titanium  
carbide.

25          39. The catheter of claim 1, further comprising a  
guide wire lumen extending through the catheter shaft.

            40. The catheter of claim 2, wherein the cathode  
comprises silicon and the x-ray unit further comprises a  
grid proximate the cathode.

30          41. The catheter of claim 2, wherein the cathode  
comprises silicon needles.

            42. The catheter of claim 11, wherein the x-ray  
unit irradiates tissue at a rate of at least about 1 gray  
per minute.

35          43. The catheter of claim 1, wherein the anode is  
coupled to a wall of the insulator, wherein the wall is  
tapered towards the anode.

            44. The catheter of claim 3, wherein:

1                   the coaxial cable comprises an outer conductor  
and a central conductor;

                  the insulator has a tubular portion with  
proximal and distal ends, the coaxial cable being coupled to  
5   the proximal end, the anode being coupled to the proximal  
end and to the central conductor of the coaxial cable, and  
the cathode being coupled to the distal end;

                  the catheter further comprises a conductive  
surface surrounding the tubular insulator, coupling the  
10   cathode to the outer conductor of the coaxial cable; and

                  the insulator and cathode define an annular  
region proximate the coupling between the cathode and the  
insulator, the annular region being screened from an  
electrical field generated between the anode and the cathode  
15   by the conductive surface and a portion of the cathode.

                  45. The catheter of claim 44, wherein the insulator  
comprises a wall depending from the proximal end of the  
tubular portion, the wall being angled toward the anode and  
the vacuum chamber.

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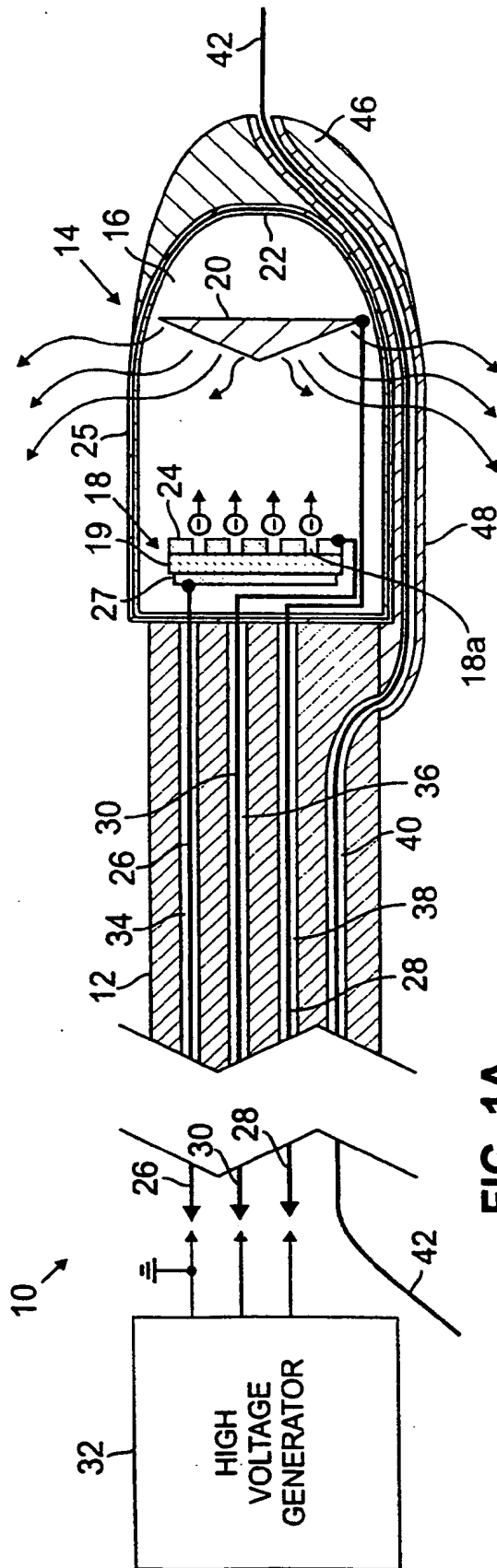


FIG. 1A

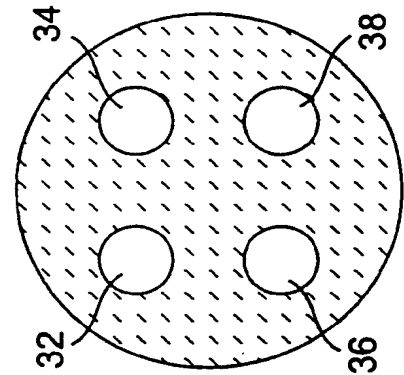
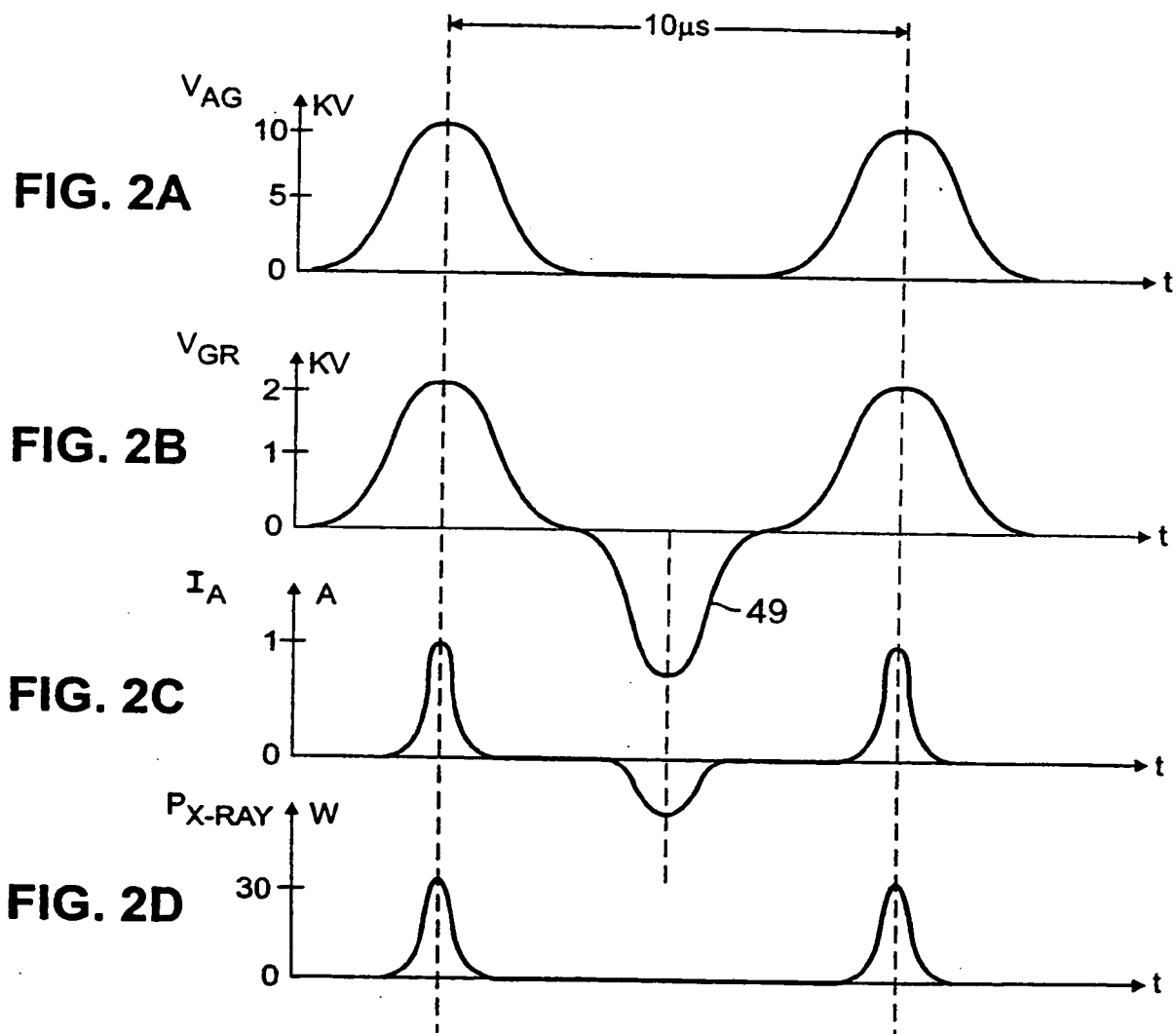
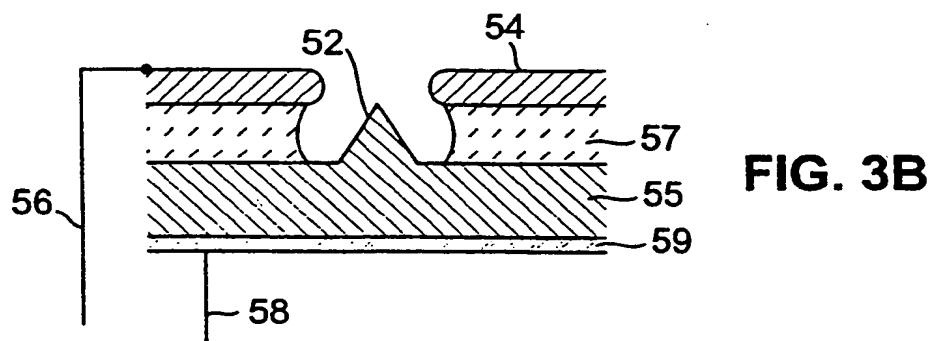
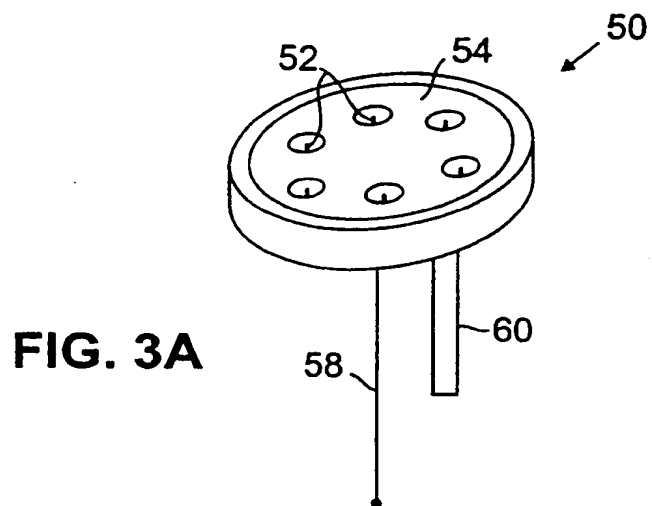


FIG. 1B

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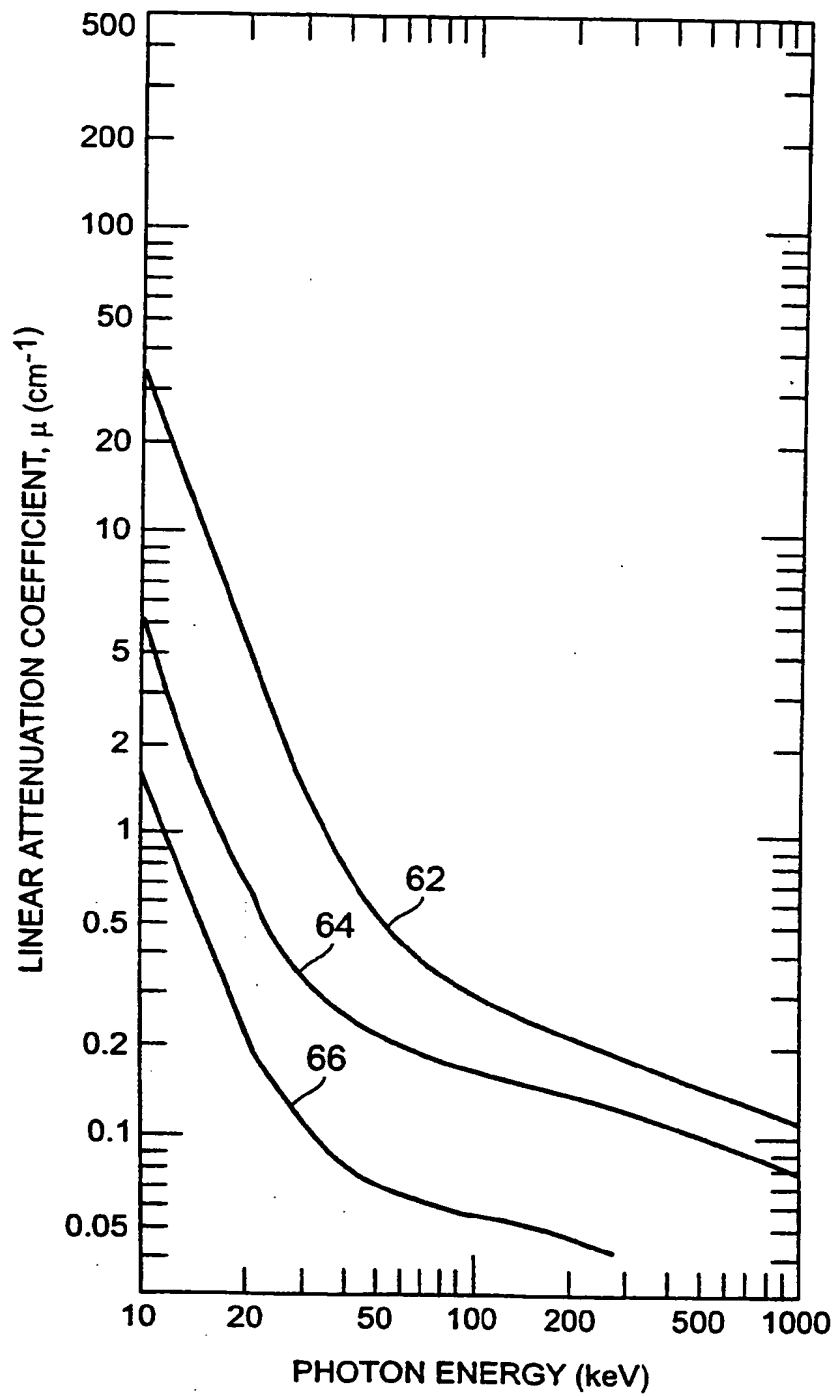
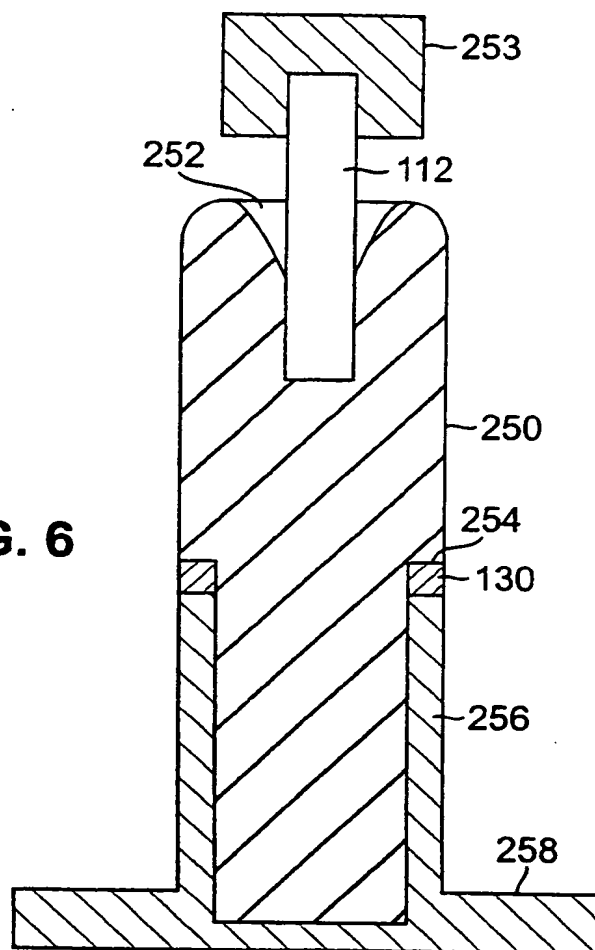


FIG. 4



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FIG. 6



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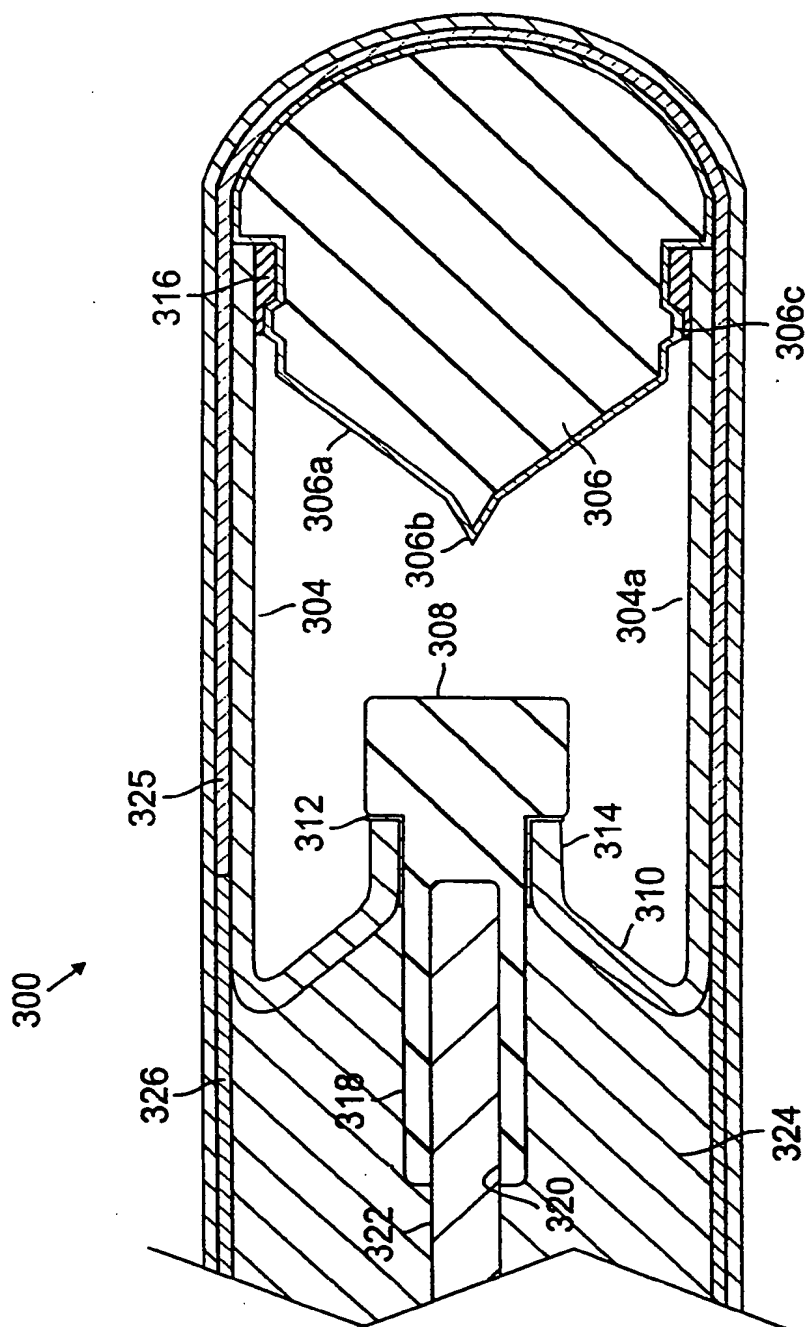


FIG. 7

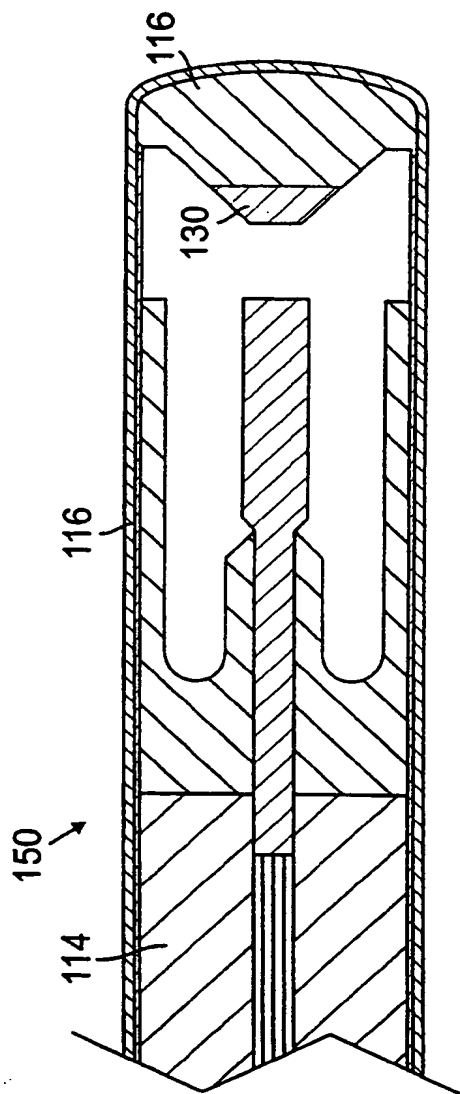


FIG. 8

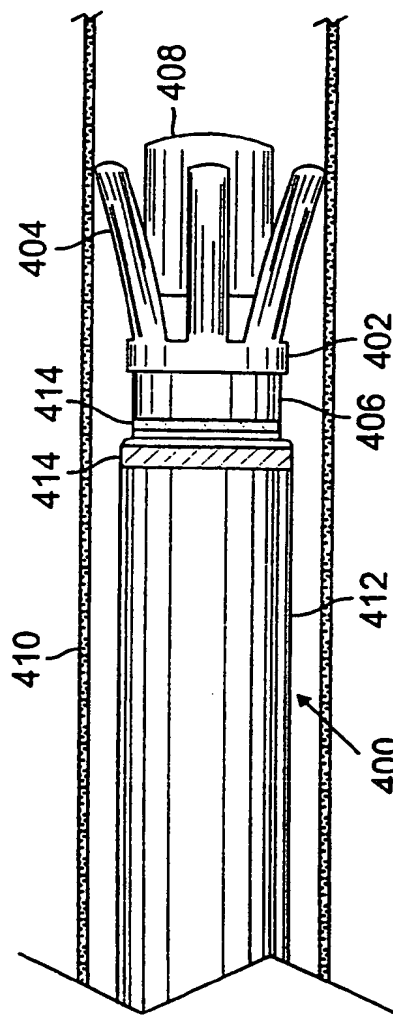


FIG. 9



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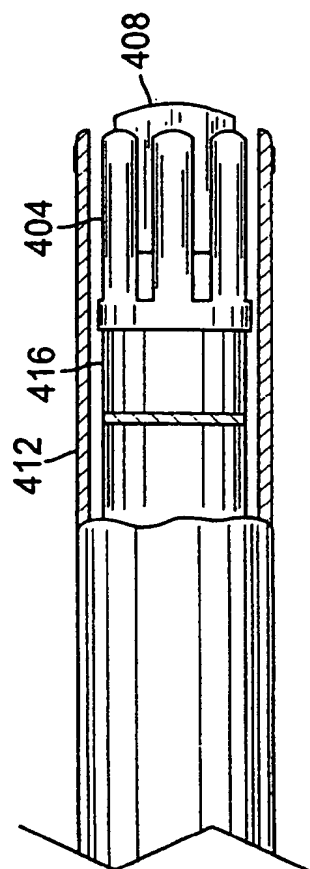


FIG. 10

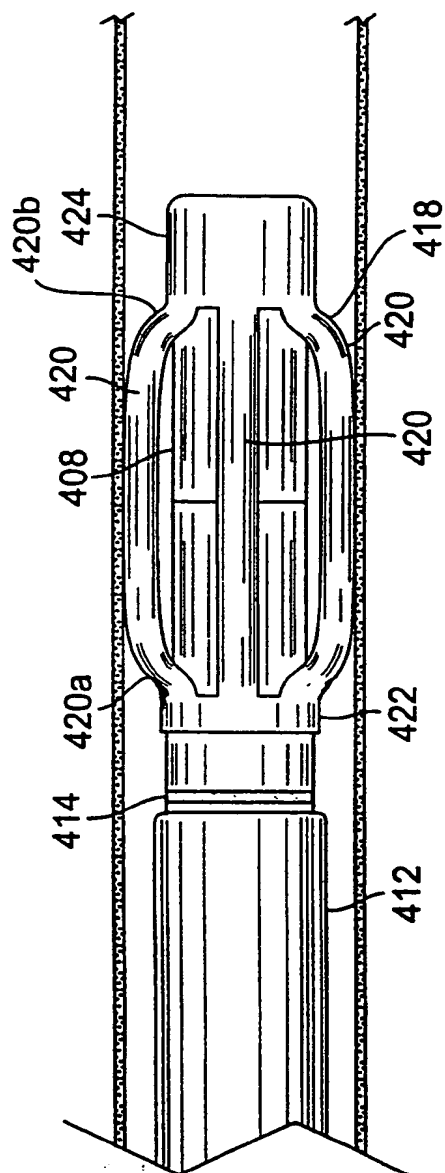


FIG. 11

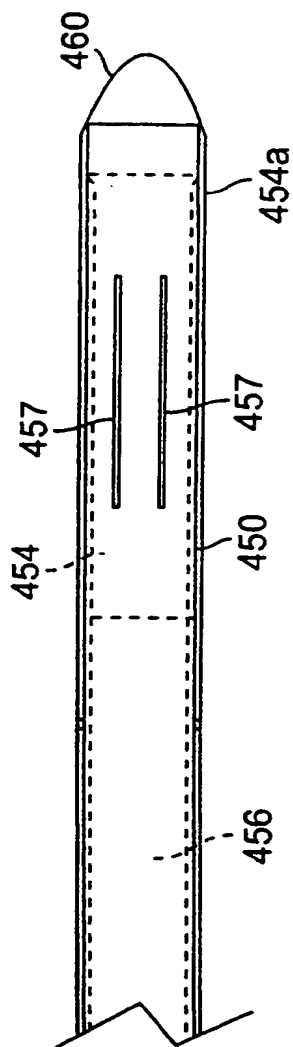


FIG. 12

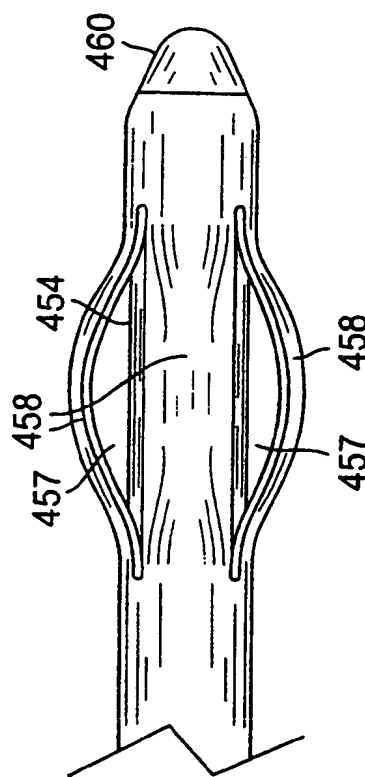


FIG. 13

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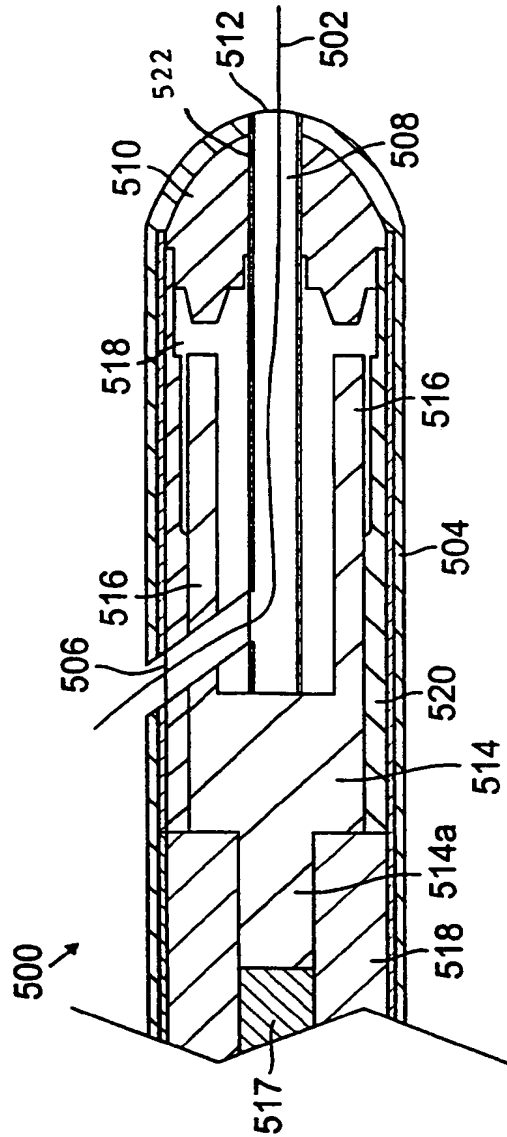


FIG. 14

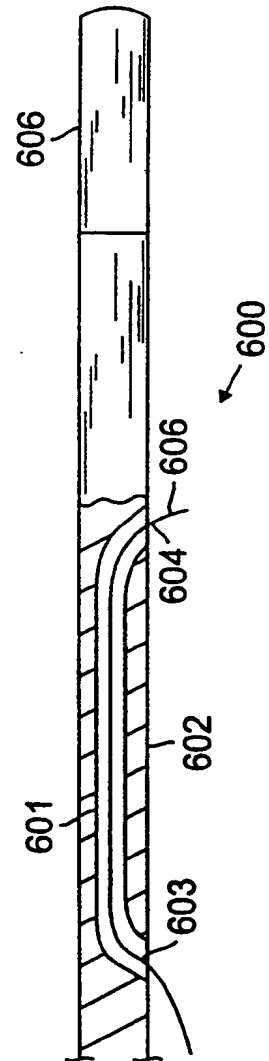


FIG. 15

# INTERNATIONAL SEARCH REPORT

International application No.  
PCT/US96/13629

## A. CLASSIFICATION OF SUBJECT MATTER

IPC(6) : A61B 6/00; A61N 5/00

US CL : 128/653.1, 659

According to International Patent Classification (IPC) or to both national classification and IPC

## B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

U.S. : 128/653.1, 659; 600/109; 606/1, 2, 7, 32; 607/804

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

APS

## C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US 5,199,939 A (DAKE et al) 06 April 1993, entire document.	1
A,P	US 5,503,613 A (WEINBERGER) 02 APRIL 1996.	1
A	SU 814-331 A (A MED ONOLOGY RES) 23 March 1981.	1-45
A	DT 2054 738 A (LOMMATZSCH et al) 10 May 1972.	1-45

☐ Further documents are listed in the continuation of Box C. ☐ See patent family annex.

* "A"	Special categories of cited documents: document defining the general state of the art which is not considered to be of particular relevance	"T"	later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"E"	earlier document published on or after the international filing date	"X"	document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
"L"	document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)	"Y"	document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art
"O"	document referring to an oral disclosure, use, exhibition or other means	"A"	document member of the same patent family
"P"	document published prior to the international filing date but later than the priority date claimed		

Date of the actual completion of the international search

11 NOVEMBER 1996

Date of mailing of the international search report

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